

HIGH EFFICIENCY PLANAR OPEN MAGNET MRI SYSTEM STRUCTURED AND
ARRANGED TO PROVIDE ORTHOGONAL FERROREFRACTORY EFFECT

FIELD OF THE INVENTION

This invention relates magnetic resonance imaging (MRI) systems, to magnet systems for producing a homogeneous imaging field for MRI and, particularly, to open magnet systems providing a remote region of field
5 homogeneity along with planar gradient coils for delivering gradient fields for spatial encoding in a remote target field region. More particularly, this invention relates to MRI magnet systems having a pair of current loops wound about a substantially planar ferromagnetic core, the core preferably including an orthogonal end piece.

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BACKGROUND OF THE INVENTION

There are known whole body MRI magnets (super-conductive, resistive iron core magnets, and permanent magnets), which produce the background B_0 field used in MRI. The useable imaging volume in these magnets is in the
15 region where the field is an extremum and provides a region of substantial field homogeneity. This volume is located in the air space centrally located between field sources. Thus, typically, MRI magnets are designed to provide a homogeneous magnetic field in an internal region within the magnet, e.g., in the air space of a large central bore of a solenoid or in the air gap between the
20 magnetic poles of a C-type magnet. A patient or object to be imaged is positioned in the homogeneous field region located in such air space. In addition to the main or primary magnet that provides the background magnetic

field B_0 , the MRI system typically has gradient and rf coils, which are, used respectively for spatial encoding and exciting/detecting the nuclei for imaging. These gradient field and rf coils are typically located external to the patient inside the geometry of the B_0 magnet surrounding the central air space.

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Prior art electromagnets such as described by Watson et al and Müller et al. and other prior art iron core magnets typically have a structural design to provide a maximum magnetic field strength at a large central air space. In addition, those types of the prior art magnets, of the iron core electro- or
10 permanent type, have a substantial edge fringe field effect, which makes it difficult to image beginning immediately at the magnet edge or even proximal to the edge of the magnet due to lack of sufficient field homogeneity.

In U.S. 5,049,848 a magnet configuration for MRI mammography is
15 disclosed. The magnetic structure 50 has a rectangular shaped magnet with at least two parallel magnetic source 5,6 connected by a ferromagnetic core flux path defining an air gap for imaging. A remote shimming C-shaped magnetic source is preferably used for shimming to decrease the front edge fringe effect of the magnetic structure 50 to create a relatively homogeneous field in the air gap
20 beginning at the front edge for effective imaging.

Solenoidal MRI magnets (superconductive, resistive) as well as iron core C and E shape electromagnets or permanent magnets are known for imaging of the whole body and its extremities. However, such whole body MRI magnets
25 are not generally well-suited for treatment of the patient with other modalities

or for minimally invasive surgical procedures guided by real time MRI because of the limited access of the surgeon to the patient. This limited access results from the field producing means surrounding the imaging volume.

Electromagnets of the C or E type iron core configuration have been designed to offer a partially open access to the patient, however, the access is still very limited with typical air gaps of only 40 cm between the pole pieces of a C type magnet. U.S. 5,378,988 describes a MRI system, which can provide access for a surgeon or other medical personnel, using a plurality of C-shape solenoidal magnets oriented to form an imaging volume in a central region of the magnets.

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Another type of magnet specifically designed for interventional surgical guidance is General Electric's Magnetic Resonance Therapy device, which consists of two superconducting coils in a Helmholtz coil type arrangement. (See US 5,677,630) The air gap for this commercial magnet is 58 cm, which typically permits access by one surgeon.

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None of those prior art magnets or MRI systems is ideal with regard to simultaneously offering real time imaging and fully open access to the patient. Many surgical procedures require three or more surgeons together with an array of supporting equipment and, thus, a fully open magnet configuration for a MRI system for interventional procedures is desirable. In addition, such open magnet configuration is desirable for patients that have claustrophobia.

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Applications other than MRI have used magnets that produce a useful field region outside the magnet geometry. U.S. Patent No. 4,350,955 describes

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means for producing a cylindrically symmetric annular volume of a homogeneous magnetic field remote from the source of the field. Two equal field sources are arranged axially so that the axial components of the fields from the two sources are opposed, producing a region near and in the plane perpendicular to the axis and midway between the sources where the radial component of the field goes through a maximum. A region of relative homogeneity of the radial component of the background field B_r may be found near the maximum. The large radial field is generally denoted as the B_0 background field in MRI applications. See also, *J. Mag. Resonance* 1980, 41:400-5; *J. Mag. Resonance* 1980, 41:406-10; *J. Mag. Resonance* 1980, 41:411-21. Thus, two coils producing magnetic fields having opposing direction are positioned axially in a spaced relationship to produce a relatively homogeneous toroidal magnet field region in a plane between the magnets and perpendicular to the axis of cylindrical symmetry. This technology has been used to provide spectroscopic information for oil well logging but has not been used for imaging.

U.S. 5,572,132 describes a magnetic resonance imaging (MRI) probe having an external background magnetic field B_0 . The probe has a primary magnet having a longitudinal axis and an external surface extending in the axial direction and a rf coil surrounding and proximal to the surface. The magnet provides a longitudinal axially directed field component B_z having an external region of substantial homogeneity proximal to the surface. Comparing this magnet geometry to that of U.S. 4,350,955, it has a background B_0 field with a cylindrically symmetrical region of homogeneity. However, this magnet

described in the copending application provides such a field in the axial or z direction (i.e., longitudinal axis direction) whereas the other provides a background B_0 field in the radial or r direction (i.e., radial direction). Preferably, the B_0 field is provided by two magnets spaced axially and in axial alignment in the same orientation and wherein said region of homogeneity intersects a plane that is located between the magnets and that is perpendicular to the axis. For MR imaging, surrounding the primary magnet are r-, z- and ϕ - gradient coils to provide spatial encoding fields.

10 It is desirable to have new and better devices and techniques for biomedical MRI applications such as open magnet MRI systems for imaging while performing surgery or other treatments on patients or for imaging patients that have claustrophobia. It is also desirable to have portable devices and imaging techniques that could be applied to a wide variety of imaging uses.

15 U.S. Patent No. 5,744,960 describes a planar MRI system having an open magnet configuration comprising two pairs of planar pole pieces that produces a magnetic field having a substantial remote region of homogeneity.

20 U.S. Patent No. 5,914,600 describes an open solenoidal magnet configuration comprises a pair of primary solenoidal coils and, located within the primary coil geometry, a bias coil system, the coils emitting an additive flux in the imaging region to generate a resulting field which provides a remote region of substantial field homogeneity.

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U.S. Patent No. 6,002,255 describes an open, planar MRI system having an open magnet configuration including a planar active shimming coil array that produces a magnetic field having a substantial remote region of homogeneity. The MRI system also includes spatial encoding gradient coils and
5 a rf coil, each preferably having a planar configuration.

The prior art magnet configurations that provide a primary background magnetic field having a remote region of substantial field homogeneity typically comprise a primary magnet system having spaced primary field emission
10 surfaces and, located between the spaced field emission surfaces, a bias magnet system having spaced bias field emission surfaces that emit an additive flux in the imaging region to generate a resulting field which provides a remote region of substantial field homogeneity. The spaced primary field emission surfaces typically are the pole pieces of a primary magnet or a solenoidal magnet facing
15 the target region.

However, a fundamental problem for obtaining an open magnet having maximal accessibility for a surgeon to conduct MRI guided surgery results from the fact that conventional magnet systems exhibit a substantial drop in magnet
20 efficiency when providing an open volume of magnetic field that is large enough for surgery to be conducted therein. Conventional iron core C-type magnet configurations provide a target field volume between co-axial pole pieces. That type of magnet configuration, even with air gap enlargement (reducing magnet efficiency), still has limited accessibility for MRI guided surgery. MRI magnet
25 systems in the form of one side "pancake type" magnet configurations (e.g. U.S.

5,331,282) generally have a set of coaxial circular coils with alternating polarity and axially shifted positions and provide relatively low level of remoteness and require a large diameter magnet for adequate field strength, thus, inhibiting accessibility to the region of field homogeneity. So-called "open" solenoid
5 superconductive magnets (e.g., U.S. 5,677,630) provide better accessibility and larger field of view (FOV) but accessibility still is limited by axial distance between two solenoidal magnets (which typically is about the width of a person's shoulders). The planar open magnet systems mentioned above (e.g., U.S. 5,378,988; U.S. 5,744,960; U.S. 5,914,600 and U.S. 6,002,255) provide
10 complete openness for excellent accessibility but suffer still from a limitation in magnet efficiency.

Thus, there remains a need to provide a more efficient, economical, open magnet MRI system having a remote region of substantial field homogeneity.

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SUMMARY OF THE INVENTION

The present inventor has discovered the magnetic field provided by a pair of spaced parallel current wires can be refracted and magnified substantially by
20 a planar ferromagnetic core that is structured and arranged to provide a substantial orthogonal ferrorefraction effect, which provides the effect of generating a set of mirror image current wires of the same polarity. It has been found that a pair of current wires spaced apart along the longitudinal axis on one side of a flat ferromagnetic core provides a field on that side of the core
25 having a distinct maximum at a remote position perpendicular to the core and

between the current wires. The position of the maximum from the core is a function of the separation between the current wires. The magnetic coupling of the current loops (providing the current wires) with a ferromagnetic core of high permeability provides an open planar magnet configuration having increased
5 magnet efficiency.

The present invention provides a planar MRI system having an open magnet configuration that produces a magnetic field having a substantial remote region of homogeneity. The MRI system includes spatial encoding
10 gradient coils and a rf coil, each preferably having a planar configuration. The magnet configuration comprises a substantially planar ferromagnetic core around which is wound a pair of primary coils, the coils providing primary current wires on one side of the planar core emitting a flux in the imaging region to generate a resulting magnetic field which provides a remote region of
15 substantial field homogeneity.

The ferromagnetic core preferably is a plate of a material having high permeability. Further, the core preferably is made by laminating layers of ferromagnetic material to form the substantially planar structure of the core.
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The planar ferromagnetic core provides a magnetic shield between the primary current wires on one side of the core plane and the return wires of the current loops on the opposite side of the core plane. In addition, the planar ferromagnetic core provides a refractive effect that magnifies the magnetic field
25 provided by the current wires by a mirror image effect when compared to the

field provided by the current wires alone. Preferably, the distance between the primary current wires emitting the field and the return wires of the current loops is minimized to minimize the amount of copper and the loss effects for the coil having a particular length of current wire on the surface of the planar core facing the target volume having a region of substantial field homogeneity. In other words the thickness of the planar core preferably is minimized. However, it also is preferred that the core material be near saturation (but not saturated) when the magnet system is operated. Thus, the core thickness is designed preferably to prevent saturation while operating near saturation to obtain a maximum magnetic induction (i.e., B-field) in the core for the ampere turns provided.

In a preferred embodiment of the invention, opposite ends of the planar ferromagnetic core are extended substantially perpendicular to the plane of the core (as defined by the portion of the core lying between the pair of current loops) on the side of the core on which lies the desired remote field of substantial homogeneity. The perpendicular extension of the ends provides an additional magnification effect of the primary current wires to the resulting field strength in the remote region of substantial field homogeneity. More preferably, the ends are extended at an acute angle with the plane of the core lying between the pair of current loops. The angle of the end extensions measured from the perpendicular is about 30° or less. Preferably, the angle of the end extensions measured from the perpendicular is about 20° or less. Most preferably, the angle of the end extensions measured from the perpendicular is in the range of about 15 to 20°.

Preferably, the end extensions also are made by laminating layers of ferromagnetic material, similar to the construction of the planar ferromagnetic core.

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In other preferred embodiments, more than one pair of primary current wires is used to provide the desired remote region of substantial field homogeneity. In addition, one or more pair of shimming current wires are used to compensate for deflections in the field profile provided by the primary current
10 wires. The shimming current wires preferably can be located on a plane that is closer to the remote region of substantial field homogeneity than the plane of the primary current wires. The use of a second substantially planar ferromagnetic core (which is much thinner than the primary ferromagnetic core and can be in sections corresponding to the location of particular shimming
15 current wires) also is preferred for the shimming current wires to obtain magnet efficiency.

The magnet configuration of the present invention can be used to provide two back to back MRI systems because the ferromagnetic core provides
20 shielding of one system from the other. Thus, a further economy can be obtained where such a back to back configuration can be utilized to provide double patient throughput such as, for example, in mass screening applications such as mammography screening.

Preferably, the magnet configuration providing the remote region of substantial field homogeneity is located on one side of a plane, which is parallel to the plane of the electromagnet ferromagnetic core and separates a patient or body component to be imaged from the magnet configuration, thereby providing
5 a planar open magnet configuration. As described herein, the background field has a direction substantially parallel to the z-axis in a rectangular coordinate system, the y-axis and z-axis define the orientation of the planar surface, and the x-axis is perpendicular to the planar surface. The z-axis may sometimes herein be called the longitudinal axis.

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As used herein, the term "remote" means that the field region is located external to the plane of current wires producing the magnetic field. As used herein, the terms "substantial homogeneity", "substantial field homogeneity" or "substantial relative field homogeneity" refer to and mean a region having
15 magnetic field homogeneity sufficient for producing MRI images of the desired quality.

As used herein, the term "substantially planar" means that the surface of the core facing the desired magnetic field is a flat surface. It should be noted
20 that the thickness of the core will depend upon the desired magnitude of the field to be provided by the magnet, i.e., upon the number of ampere-turns used to excite a given field.

The distance of the region of homogeneity from the planar surface can be
25 controlled by varying the spacing between the primary coil pairs. The primary

coil pairs are the major source for background field and determine the basic field strength. The shimming coil pairs provide compensation for field deflection resulting from the primary coils and contribute to the degree of homogeneity of the background field. The size, geometry and spacing of the coils can influence the size of the homogeneous region.

The MRI system also preferably has (i) a planar xyz gradient coil system that produces a gradient field for spatial encoding in the region of homogeneity, i.e., the imaging volume or target region, as well as (ii) rf excitation and receiving coil (or coils).

The size of the magnet configuration and MRI system in accord with the present invention can be varied to provide whole body imaging or portable systems for localized imaging.

BRIEF DESCRIPTION OF THE DRAWINGS

FIG. 1A is a cross sectional view of an isometric illustration of a magnet configuration in accord with one embodiment of the present invention.

FIG. 1B is an illustration of the background magnetic field B_0 along the x-axis provided by the magnetic configuration of FIG. 1A.

FIG. 2A is a cross sectional view of an isometric illustration of a magnet configuration illustrating one pair of current loops coupled with a planar ferromagnetic core in accord with the present invention.

FIG. 2B is an equivalent magnetic circuit for the magnet configuration of FIG. 2A.

FIG. 2C is an illustration of mirror image refraction magnification effect provided to the field generated by a current wire by two perpendicular ferromagnetic plates positioned adjacent the length of the current wire.

FIG. 2D is a graph of the maximum field strength B_{\max} measured at a
5 point along the x-axis, which is perpendicular to the core surface, for one pair of coils around a planar core (FIG. 2A) having variable increments of thickness (t_c).

FIG. 3 is a cross sectional view of an isometric illustration, in part, of a magnet configuration for a double MRI system in accord with an embodiment of
10 the present invention.

FIG. 4 is an illustration of the longitudinal axial positioning of coils for x-z (2 dimensional) shimming of the remote region of substantial field homogeneity.

FIG. 5A is an illustration of an experimental device used to measure field
15 strength for magnet systems in accord with the present invention.

FIG. 5B is graph showing the relationship of maximum field strength and the length of ferromagnetic end plate returns positioned orthogonal to the ferromagnetic core plate.

FIG. 5C is a graph illustrating the maximum field strength for the device
20 of FIG. 5A with and without ferromagnetic end plate returns, showing the increased field strength due to the refraction effect provided by the present invention.

FIG. 6 is a graph illustrating the affect of the angle of the ferromagnetic end plate returns on the maximum field strength.

FIG. 7 is a schematic illustrating a planar remote field x-gradient coil in the form of an array of current loops and further illustrating at the center of each loop the vector magnetic dipole associated with the loop. Also illustrated are the field vectors generated by the magnet dipoles providing an

5 approximately constant gradient field.

FIG. 8A is a schematic plan view illustration of a planar remote field x-gradient coil in the form of an array of current wires.

FIG. 8B is a cross sectional view of the x-gradient coil of FIG. 8A along the z-axis illustrating field lines and further schematically illustrating the B_z

10 field as a function of x at $y=z=0$ showing an approximately constant gradient field with respect to x.

FIG. 8C illustrates a simple three current wire array and a simple computation vector diagram to aid the understanding of an X gradient field with the current wire array.

FIG. 9A is a schematic plan view illustration of a planar remote field y-gradient coil in the form of a current wire array.

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FIG. 9B is an isometric illustration of the planar array of FIG. 9A including an illustration of the remote y-gradient field provided by the array of FIG. 9A.

FIG. 9C is an illustration of the y-gradient field profile along the y-axis of Fig. 9B.

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FIG. 9D is an illustration showing a simple six (6) wire array for a y-gradient coil for analysis of coil capability.

FIG. 10A is a schematic plan view illustration of a planar remote field z-gradient coil in the form of a rectangular loop array.

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FIG. 10B is a schematic cross sectional view in the xz plane of the wires of the loop array of FIG. 10A and illustrates the remote z-gradient field provided by the z-gradient coil.

FIG. 11A is a schematic illustration of three rf coils for excitation and
5 detection of the nuclear magnetic resonance signal, which provide a remote x-directional rf field orthogonal to the z directional background field.

FIG. 11B illustrates the x-directional field provided by the rf coil of FIG. 11A which is orthogonal to the z-axis.

FIG. 12 is a front elevational view of two magnet configurations of the
10 present invention in a "face to face" arrangement that provides a field having a maximum approximately twice that of one configuration standing alone.

FIG. 13A is an illustration of part of a magnet configuration in accord with a preferred embodiment of the invention showing a construction to provide a layer operating near magnetic saturation adjacent to the current wires
15 providing the primary magnetic field.

FIG. 13B is an equivalent magnetic circuit for the magnet configuration of FIG. 13A.

FIG. 14A is an illustration of a stepped end plate extension for providing orthogonal refraction in accord with one embodiment of the present invention.

20 FIG. 14B is an illustration of an alternative construction for a stepped end plate extension for providing orthogonal refraction in accord with another embodiment of the present invention.

DETAILED DESCRIPTION OF THE INVENTION

The invention will be described with reference to the drawings wherein like reference numerals in different figures refer to the same component. In accord with the present invention, a magnet configuration for a MRI system is
5 an open device that provides a background B_0 magnetic field having a remote region of substantial homogeneity and oriented substantially parallel to the plane of the ferromagnetic core portion that is between a primary pair of current wires that provides the magnetic field.

10 As illustrated in FIG. 1A, one embodiment of a magnet system 10 in accord with the present invention has a pair of primary current wires dived into two portions. A first portion of primary current wires 21 (provided by N_2 turns of the coil), and a second portion of primary current wires 22 (provided by N_1 turns) on one side of a ferromagnetic plate 25 (around which the coils are
15 wound). Each portion of primary current wires have a first group of wires having one polarity spaced apart from a second group of wires having same polarity along the longitudinal axis (i.e., Z axis in FIG. 1A) of the ferromagnetic plate, which serves as a core for the coils. In addition, the ferromagnetic plate provides high magnetic conductivity (due to high permeability of the material) in
20 the core region between the primary current wires and the return wires provided by the loops, which return wires have opposite polarity. The ferromagnetic plate also can be considered to provide a magnetic shielding effect between the primary current wires lying on one side of the plate and the return wires of the coils lying on the opposite side of the plate (which can be

used to provide independent magnetic field generation on opposite sides of the core plate).

Preferably, ends 26 of the ferromagnetic plate 25 are extended orthogonal
5 to the longitudinal axis (i.e., in the X direction in FIG. 1A). The ends 26 also preferably can have a pole piece 27, which can have a geometry to aid in shaping the magnetic field provided by the magnet system 10. However, the primary field is generated by the pairs of primary current wires.

10 Shimming of the magnetic field is provided by a first pair of shimming current wire groups 31 having N_{s1} wires in each group and a second pair of shimming current wire groups 32 having N_{s2} wires in each group. In this embodiment, the current in the first pair of shimming current wires 31 has opposite polarity to the second pair of shimming current wires 32. However,
15 the polarity can be in the same direction in both pair shimming wires. The pairs of shimming wire groups are wound about a ferromagnetic plate 35 and spaced along the longitudinal axis thereof. Preferably, ferromagnetic plate 35 can be segmented to maximize the shimming field contribution from each shimming current wire and, at the same time, to minimize a shunting effect on
20 the primary field. Orthogonal end extensions and optionally even pole pieces, similar to plate 25, may be provided in certain circumstances. Ferromagnetic plates 25 and 35 are positioned in parallel. The primary and shimming current wires are spaced symmetrically about a centerline of the ferromagnetic plates in the X direction (i.e., $X=0$), respectively. Typically, the centers of the core plates
25 are aligned with the X,Y plane as illustrated.

It should be noted that the shimming of the target field region can be either active or passive.

5 The magnetic field in the Z direction B_z along the X axis that is provided by the magnet system 10 (FIG. 1A) is illustrated in FIG. 1B. Curve B_1 is the field provided by primary current wires 21. Curve B_2 is the field provided by primary current wires 22. Curve B_{s1} is the field provided by shimming current wires 31. Curve B_{s2} is the field provided by shimming current wires 32. Curve
10 $B(N_1 + N_2)$ is the sum of the fields provided by the primary current wire group pairs ($B_1 + B_2$). Curve $B(N_1 + N_2 + N_{s1} + N_{s2})$ is the sum of the fields provided by the primary and shimming current wire group pairs. T designates the region of substantial field homogeneity of magnetic field provided by the magnet system.

15 The ferromagnetic plates preferably are made by laminating together sheets of ferromagnetic material having the desired shape, e.g. the shape of a longitudinal cross section of the plate and optionally including orthogonal end extensions. As in transformer construction, the use of laminated sheets reduces induced currents (which can arise due to pulsing of gradient coils in a
20 MRI system) and provides a more technological capability in manufacturing the magnet.

FIG. 2A illustrates a magnet system 50 having only one pair of current wire groups 51 and a ferromagnetic plate 55 that consists of a plurality of
25 sheets of ferromagnetic material. The thickness of the ferromagnetic plate 55 is

variable depending upon the number of sheets laminated together to form the plate. A magnet system as illustrated was constructed to determine the effect of a ferromagnetic plate of thickness t_c on the shielding of the return wires and the refraction effect. The coils providing the current wire groups were made of 18 gauge copper wire with 100 turns per coil and were spaced 8 cm apart on a core plate 30 cm long and 24 cm wide. The field was measured perpendicular to the plate at the center of the two coils with a current of 3 amps in each coil. The thickness of the plate t_c was varied in increments of 0.3 mm, the thickness of the sheets of M6 transformer steel used for the laminations.

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An equivalent lumped parameter magnetic circuit for the magnet system 50 of FIG. 2A is illustrated in FIG. 2B. The equivalent circuit can be used to compute a minimum thickness of the ferromagnetic core plate providing a maximum in the remote field. This computation is made to differentiate a contribution generated by the current wires and the contribution due to the mirror image effect due to refraction by the orthogonal core plate structure. Computations using the equivalent circuit can be performed by conventional circuit analysis well known to those skilled in the art.

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FIG. 2D is a graph illustrating the maximum field strength of the coils as a function of the number of laminated sheets. The fall off at 18 laminations (i.e., sheets) is due to an artifact caused by low saturation because of the low current, which was kept constant for the experiment.

The quadro mirror effect provided for a magnetic field generated by a straight wire I_1 by mutual orthogonal ferromagnetic core plates of infinite length is illustrated in FIG. 2C. The presence of the mutual orthogonal ferromagnetic plates has the effect of inducing a field from a unipolar current wire pair where the field can be calculated as if including a contribution from three other imaginary wire pairs (I_{2i} , I_{3i} , I_{4i}) located on opposite sides of the mutually orthogonal ferromagnetic plates from the actual current wire I_1 as shown in the figure. A pair of wires I_1 produce a magnetic field B_{1m} along the X axis. The orthogonal ferromagnetic end plates at each end of the magnet configuration provide an orthogonal equipotential boundary condition at high permeability and provide an effect to magnify the magnetic field produced by the actual pair of current wires I_1 alone thereby providing a total magnetic field B_{tot} that is equal to the sum of magnetic fields produced as if imaginary wires I_{2i} , I_{3i} , I_{4i} were actual wires considered without the ferromagnetic core boundaries (this total field is called the "ferrorefraction" effect herein). The total magnetic field B_{tot} produced by wires I_1 and the orthogonal ferromagnetic core plates is equal to the sum of B_{1m} , B_{2im} , B_{3im} and B_{4im} . In actual magnet systems where the plates cannot be infinite for practical reasons, the magnification effect is less than theoretical.

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Because of the magnetic shielding provided by the ferromagnetic core plate, the magnet structure of the present invention can be used to obtain two individual "back to back" magnet systems wherein the current to the primary coils is used to operate both magnet systems (i.e., providing a "double sided" magnet system). This can provide a cost effective operation mode for

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applications where back to back systems can be utilized, for example, in mass screening applications such as, specifically, mammography screening. FIG. 3 illustrates an embodiment of such a magnet structure. The back to back magnet system 60 has a first pair of primary coils 61 having N_m turns, and a
5 second pair of coils 62 having N_1 turns coupled with a ferromagnetic core plate 65. The primary coils are spaced along the longitudinal axis (i.e., Z axis).

Ends 66 of the ferromagnetic core plate 65 are extended orthogonal to the longitudinal axis (i.e., in the X direction), both to the right and to the left of
10 plane of the core plate 65. The ends 66 also preferably have a pole pieces 67r and 67 , which can have a geometry to aid in shaping the magnetic field provided by the magnet system.

The double side magnet is provided with symmetrical primary current
15 wire pairs and shimming current wire pairs for the right and left sides. Shimming of the magnetic field produced by the primary coils is provided on both sides of the plate 65; on the right side by a first pair of shimming coils 71r having N_{s2r} turns and a second pair of shimming coils 72r having N_{s1r} turns, and on the left side by a first pair of shimming coils 71 having N_{s2} turns and a
20 second pair of shimming coils 72 having N_{s1} turns. The pairs of shimming coils are wound about a ferromagnetic shimming core plate 75r or 75 , respectively, and spaced along the longitudinal axis thereof. Ferromagnetic shimming core plate 75r or 75 also can have orthogonal end extensions and pole pieces, similar to plate 25 (in FIG. 1A). Ferromagnetic shimming core
25 plates 75r and 75 are positioned in parallel with core plate 65. The primary

and shimming coils are spaced symmetrically about the longitudinal centers of the ferromagnetic core plates, respectively. Typically, the longitudinal centers of the plates are aligned with the X,Y plane as illustrated. The background magnetic field B_0 in the Z direction along the X axis that is provided by the right side of the magnet system is illustrated. A field having the same magnitude and opposite polarity is generated on the left side.

An alternative shimming scheme for two dimensional shimming is illustrated in FIG. 4. For illustration purposes, only the positions of the coils along the Z axis are shown. The shimming coil system 80 provides shimming of the B_z field in both the X and Z spatial directions. Two pairs of shimming coils 81, 82 having different numbers of wire turns is centered about the X axis and provides a maximum shimming field along axis X_1 where in this case $Z = 0$. Two pairs of shimming coils 83, 84 are shifted along the Z axis in the positive direction and centered about axis X_2 . Another two pairs of shimming coils 85, 86 are shifted along the Z axis in the negative direction and centered about axis X_3 . Each of the three sets of coils (81,82; 83,84 and 85,86) are designed to provide the same X direction shimming effect. By shifting the position of the symmetric sets of coil pairs along the Z axis, a shimming effect also is provided in the Z direction. Thus, a region of the magnetic field denoted by T in the X and Z directions is affected by the three sets of shimming coils.

A device used to measure the magnetic field provided by primary electromagnetic current wires and a planar core is illustrated in FIG. 5A. Two pair of primary coils were used. In the first pair, each coil had 100 turns of 18

gauge copper wire and a coil width t_N along the Z-axis of about 2.25 cm. Each coil was positioned along the Z-axis at a distance h_1 from the X-axis, providing a separation of $2h_1$ between the first coil pair. In the second pair, each coil had 547 turns of 18 gauge copper wire and a coil width t_N along the Z-axis of 4.5 cm. These coils were positioned along the Z-axis at a distance h_2 from the X-axis, providing a separation of $2h_2$ between the second coil pair. The core was made of M6 transformer laminar steel having saturation inductance $B_s = 1.8$ T (Tesla) and the core had a total thickness of 0.8 cm (each lamination thickness being approximately 0.3 mm), a width along the Y-axis of 24 cm, and a variable length along the Z-axis of up to 30 cm. The orthogonal end extension is made of the same M6 transformer steel and extends 9 cm in the X-direction from the surface of the planar core plate. Shimming devices can be added, if desired.

The effect of the length of the orthogonal end plate extensions was determined by varying the length of the end plate in increments of 1.0 cm from 0 to 9.0 cm. The results are presented in the graph illustrated in FIG. 5B. In the experiment, only one pair of current wire groups each having 547 wires was used. The wire groups were positioned at $h_2 = 8$ cm. The maximum field strength was measured perpendicular to the core plate along the X-axis.

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FIG. 5C shows the maximum field strength measured experimentally for two pairs of coils as described above, $h_2 = 8$ cm, $h_1 = 6$ cm, and 3 amp current in the coils. The device of FIG. 5A was operated with and without the end plate extensions (9 cm in length). The dashed lines indicate calculated values and the solid lines indicate experimentally measured values. As can be seen,

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without the end plates the device provided a maximum field of about 80 gauss in a region where $X = 5.5$ to 7.0 cm (illustrated by solid squares connected by solid lines). This can be compared with calculated field value of 40 gauss for free space, which was calculated using the Biot-Savart law methodology

5 (illustrated by solid circles connected by dashed lines). With the end plate extension added, the device provided a maximum field strength of about 150 gauss in the region where $X = 6.5$ to 8.0 cm (illustrated by x's connected by solid lines). The comparison calculated values (illustrated by solid triangles connected by the dashed line) were made using a quadru mirror image

10 methodology. It should be noted that the end plate extension causes the region of maximum field strength to shift outward along the X axis.

Instead of orthogonal end plates, the angle of the end plates from the vertical toward the X-axis was varied to determine the affect of the angle on the

15 maximum field strength. all parameters except for the angle of the end plates was kept constant. As illustrated in FIG. 6, the maximum field strength increases with the angle of the end plate, at least up to an angle of 20° from the vertical.

20 In accord with the present invention, a MRI system comprising a magnet configuration, for example, as illustrated in FIG. 1A also comprises a set of xyz gradient coils to provide spatial encoding in the remote region of background B_0 field homogeneity for imaging. In keeping with the planar open design of the magnet geometry and configuration of the present invention, planar xyz

25 gradient coils preferably are used. Advantageously, the remote field xyz

gradient coils are preferably planar open surface coils. Preferably, the entire gradient coil is substantially in the same plane and located in front of the primary and shimming current wire pairs. Each single planar gradient coil produces a remote gradient field along the x, y or z axis in the target region of background field homogeneity.

As illustrated in FIG. 7, a planar remote field x-gradient coil can be constructed using multiple current loops in a current loop array configuration. As shown, two external loop sections are configured to have mutually opposite current polarity. Two internal current loop sections are configured also to have mutually opposite current polarity and the current polarity in the inner loop sections is also mutually opposite its corresponding external loop section (FIG. 7A). The array loops are symmetric with respect to the y and z axes.

Also illustrated on the right side of FIG. 7 is a simple field computation approximation using the magnetic moment vectors associated with the current loops. The basic B_z field components at x_1 contributed by each of the four magnetic dipoles are shown. The vector sum of the field from the two main current loops with current I_1 and turns N_1 is denoted B_{z1} and the two bias current loops with current I_2 and turns N_2 is denoted B_{z2} . Also illustrated on right hand side of FIG. 7 are (i) the vector sum of the field from the current loops which produces a gradient field $B_{z,x}$, which is substantially linear over the target field region Δx_1 and which corresponds to the region in which the background field $B_{z,o}$ (also shown) goes through an extremum.

An alternative x-gradient coil structure is illustrated in FIG. 8A, with field lines of the two opposing fluxes illustrated in FIG. 8B with the resultant x-gradient field, z component, $B_{z,x}$ which is substantially linear over Δx_1 illustrated in the right side of FIG. 8B. This coil structure is a current wire array and will provide a remote linear x-gradient field similar to the current loop array depicted in FIG. 7. FIG. 8C illustrates a simple three wire system for demonstration purposes.

The x-gradient coil structure has ampere wire distribution on a planar surface 215 with two external unidirectional current wire sections N_{11x} and N_{12x} with a central unidirectional current section N_{21x} having opposite current polarity to provide a central bias field, thereby providing a remote z-component field having substantially constant gradient as illustrated in FIG. 8B.

It should be noted that the x-gradient coil geometry is similar in concept to the magnet geometry for the B_0 background field.

As illustrated in FIG. 9A, a planar remote field y-gradient coil can be constructed using a current wire array. As illustrated, the current wire array preferably has four sections corresponding to four y,z quadrants: first having coordinates 0, -y, z; the second having coordinates 0, y, z; the third having coordinates 0, -y, -z; and the fourth having coordinates 0, y, -z; in other words, the quadrant boundaries are the y and z axes in the $x=0$ plane. Each quadrant section is divided in subsections along the y axis. The subsections at each z-level ($z_1, z_2, z_3 \dots$) are symmetrical about and have opposing current polarity

with respect to the z -axis. The subsections are also symmetrical about and have the same current polarity with respect to the y -axis. As can be seen from FIG. 9B, the number of current wires remains constant over y at each level of z , however, the number of wires having positive and negative current polarity varies along y . The current wires providing the remote y -gradient field are connected by return wires (dashed in FIG. 9A) which are located far enough away from the desired wire current array to minimize target field disturbance.

FIG. 9B illustrates the current wire array, which is located in the $x=0$ plane, extends in the y direction from $+y_0$ to $-y_0$, and provides a y -gradient field in a remote region. It illustrates only the operative current wires of the array of the wiring schematic of FIG. 9A. The y -gradient field is a field directed in the same direction z as the background field B_0 and which varies linearly with y . On the right side of FIG. 9B is an illustration of the y -gradient field at the remote plane $x=x_0$. Note that the field is sinusoidal as a function of y so that the region of linearity is localized around $y=0$. The region of linearity can be specified as a fraction of y_0 which is approximately $0.5 y_0$. Thus, increasing the spatial extent of the current wire array in y can increase the region of linearity in y .

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The gradient field depends upon y because the distribution of current wires with positive and negative currents parallel to the y axis changes with the y position. FIG. 9B illustrates the current pattern changes at $y=y_1, y_2$, etc. Thus, between y_0 and y_1 , the current pattern as a function of z is constant and then changes to a different pattern between y_1 and y_2 . The field produced by

each of these current patterns is shown as a function of x at each incremental y position where the z pattern of the current wire array changes. In each case, the field goes through an extremum at the same value of x , in this case x_0 , however, the amplitude of the extremum changes from positive to negative producing the gradient field.

FIG. 9C is an illustration of the gradient field profile. Because the field is an extremum at the remote $x=x_0$ plane, it is a region of relative homogeneity. The region of homogeneity of the gradient field extends in the x direction over a distance Δx_0 .

FIG. 9D illustrates the simplest six current wire array, which is useful as an analytical model for easy computation of the remote y -gradient field.

A planar remote field z -gradient coil can be constructed between the primary current wire pairs. FIG. 10A shows the z -gradient coil as a set of current wires a_{11} , b_1 and a_{12} at positive values of z and a symmetric set of opposite polarity current wires a_{22} , b_2 and a_{21} , located at corresponding negative values of z . The current in wires a_{11}, a_{12} have current in one direction with the bias wire set b_1 having current in the opposite direction. The wire sets a_{21}, a_{22} and b_2 on the opposite side of the y -axis have currents flowing in the opposite directions to the currents in the corresponding wire sets a_{11}, a_{12} and b_1 . A remote z -gradient field is provided in the target background field. All of the current wires in sets "a" are preferably are connected, as illustrated, to be supplied by a single power source. Similar to the field produced by the y -

gradient coil of FIG. 9A, these two sets of current wires produce a field maximum, each of opposite sign at the remote plane $x=x_0$ at different values of z . This produces the gradient field illustrated in FIG. 10B.

FIG. 10B shows a side cross sectional view of the z -gradient coil illustrating field line orientation in the xz plane at $y=0$ of the wire sets illustrated in FIG. 10A with a plot of the z component of the z -gradient field B_{zg} illustrating a substantially constant gradient over a range of z .

As illustrated in FIG. 11A, a planar rf coil set useful in connection with the present invention is illustrated, which provides a remote region of approximate rf field homogeneity wherein the major components are oriented orthogonal to the z component of the background field, i.e., oriented in the x -direction. The three coils illustrated can be used singly or in combination with each other to produce an rf field perpendicular to the B_0 field in different regions of the target volume. By varying the number of coaxial current loops in the rf coils, which can be in different but parallel planes, the region of the x -oriented rf field homogeneity can be positioned in the region of remote field homogeneity of the background field. FIG. 11B illustrates the x -directional field provided by the rf coil of FIG. 11A, which is orthogonal to the z -axis. However, it will be appreciated by those skilled in the art that the rf field must be perpendicular to the B_0 field (which for discussion in this specification is in the z direction). Thus, an rf field in the y direction can also be illustrated in a similar manner and provided used for excitation and detection of the nuclear magnetic resonance signal.

The open magnet configuration providing a remote background field for MRI can be made in any size to provide whole body capability or a portable system for more localized imaging. Shimming coils can also be used to define
5 and position the remote region of background field homogeneity.

Planar xyz gradient and rf coils preferably are positioned below the patient table in front of the primary and shim current wire pairs and can extend over the surface between end plate extensions. The particular geometry and
10 size of the gradient and rf coils can be optimized for each particular application, as is well known in the art.

Two individual magnet systems in accord with the present invention can be positioned adjacent to each other (i.e., "face to face") a distance D apart, as
15 illustrated in FIG. 12. In such case, the magnetic fields provided by the two individual magnet systems, one located on the Z_1 axis and the second on the Z_2 axis, are additive and provide a substantially increased field strength at the with a maximum along the X axis, the maximum being located on the Z_0 axis for the illustrated system. Shimming current wire pairs (not shown) preferably
20 also are used to provide the desired field shape for the application. Gradient and rf coils (not shown) are also provided for MR imaging, as discussed above.

In order to obtain the refraction effect of a planar ferromagnetic core, it has been discovered that it is sufficient to have only a thin layer at the surface
25 of the ferromagnetic plate material performing near saturation to obtain

orthogonal refraction effect, while the major portions of the core can be saturated or "oversaturated." Therefore, a preferred ferromagnetic core structure to benefit from the refraction effect and to prevent the core from saturation at surface layer adjacent to the current wire pairs emitting the remote field is illustrated in FIG. 13A. The magnet system 130 comprises at least one set of current wire pairs (one set of current wires 132 being illustrated) and a primary ferromagnetic core plate 135 with an end plate extension 135a (135a' optional for a second field, if desired). Ferromagnetic plates 136, 137 are constructed in parallel with the primary core plate 135, but have substantially thinner plate thickness. End plate extensions 136a, 137a are also constructed parallel to primary end plate extension 135a. End plate extensions 136a and 137a are connected structurally and magnetically by ferromagnetic plate 138 to form a closed loop. The other end (not shown) of the magnet system 130 is similarly constructed in a symmetric mirror image construction. Thus, plates 136 and 137 are parts of a continuous mechanically closed magnetic flux loop.

A bias coil constructed in three sections IN_{b1} , IN_{b2} and IN_{b3} generates a magnetic flux in the closed magnetic loop. The direction of the magnetic flux in plate 136 and end plate extension 136a that is generated by the bias coil is opposite to the flux generated by the primary current wire pairs 132. One or more bias coil sections are used to provide a bias flux in the magnetic loop preferably to maintain the total magnetic flux in plates 136 and end plate 136a at or near saturation of the B field. Thus, plate 136 and end plate 136a provide substantially orthogonal refraction of the primary current wire pairs and

provide independent magnetic field generation for the current wire pairs on the right of the core from the magnetic field generation of the current wire pairs on the left of the core plate 136. The bias coils provide a demagnetization flux to compensate for the saturation induction of the primary field coils and maintain
5 the surface layer near saturation.

Core plates 135, 135a, 137 and 137a typically are made of M6 transformer steel having a saturation inductance with $B_s = 1.8$ T. Preferably, core plates 136 and 136a are made of steel having a higher saturation
10 inductance with $B_s = 2.2$ to 2.5 T, as well as a higher permeability prior to saturation. An equivalent magnetic circuit for the magnet system 130 is illustrated in FIG. 13B to provide an analytical tool for design optimization. Shimming, gradient and rf coils, as discussed above, are provided for MR imaging. Optionally, core plates 136', 136a', 137' and 137a' are provided along
15 with suitable bias coils, if a "back to back" or double magnetic field system is desired.

It is known that certain magnetic properties of materials can be represented by a B vs H curve where B is the magnetic induction and H is the
20 magnetic intensity. The ferromagnetic core material preferably operates close to saturation for the material used. Typically, close to saturation means that the system is designed, structured and arranged so that the ferromagnetic core layer adjacent to the primary current wire pairs operates with the ferromagnetic material having a magnetic intensity H within about 20% of H_{sat} , more
25 preferably within 10% of H_{sat} , most preferably within 5% of H_{sat} .

Alternative stepped end plate extension structures for providing an orthogonal quadrupole ferrorefraction effect are illustrated in FIGs. 14A, 14B. In FIG. 14A, a primary ferromagnetic core plate 140 with a stepped end plate extension 140a (only one end of magnet system being illustrated) accommodates primary current wire pairs that are divided into four sections 142a, 142b, 142c and 142d. In this embodiment, the dimensions of the steps are shown to be the same for each step. Thus, some of the mirror images of the coils are overlapping, providing target field magnification. In FIG. 14B, a primary ferromagnetic core plate 150 with a stepped end plate extension 150a accommodates primary current wire pairs 152a, 152b, 152c and 152d that are located in steps having non-equal dimensions. Thus, in this embodiment, the dimensions of the steps vary for each step to obtain the most efficient overlap of the mirror images to generate the desired remote field and obtain maximum target field magnification. In FIGs. 14A and 14B, only one end of the magnet system is illustrated (compare FIG. 3). Also, a double magnet system can be constructed using the same primary coils by providing a mirror image structure on each side of the ferromagnetic core plate. Shimming coils can also be used as discussed above.

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For the general case, the ladder step end plate extension structures are designed with different dimensions for each step to shape the remote field and provide desired homogeneity for particular applications. Larger numbers of current wire pairs typically will be positioned at a larger distance from the X-

axis (i.e., larger separations along the longitudinal axis). The design also will be optimized for the particular geometric constraints.

The invention has been described in detail with reference to preferred
5 embodiments thereof. However, it will be appreciated that, upon consideration
of the present specification and drawings, those skilled in the art may make
modifications and improvements within the spirit and scope of this invention as
defined by the claims. For example, other magnet configurations can be use for
providing the primary field or for shimming. Indeed, the current wire pairs
10 generating the primary field or the shimming field can be divided into any
number of groups having different numbers of wires and different spacings
along the longitudinal axis.